Comparison of 0.8μm and 1.3μm images of optical coherence tomography using fiber-optic low-coherence interferometers

Masato Ohmi, Takayuki Kurata, Keiko Kishida, Makiko Takatsuka and Masamitsu Haruna
School of Allied Health Sciences, Faculty of Medicine, Osaka University
1-7, Yamada-Oka, Suita, Osaka 565-0871, Japan
TEL: +81-6-6879-2576 FAX: +81-6-6879-2469
E-mail: ohmi@sahs.med.osaka-u.ac.jp

Abstract: Optical coherence tomography (OCT) is very promising for high-resolution cross-sectional imaging underneath the surface of biological tissue in the medical use. In this paper, two OCT images for 0.8 and 1.3μm are compared in fiber-optic low-coherence interferometers. Our experiment suggests that the maximum light penetration depth of 1.3μm OCT is almost twice as large as that of 0.8μm OCT, while the latter can image micro tissue structures near the surface with higher contrast.

1. Introduction

Very recently, keen interest has been in development of optical coherence tomography (OCT) in the field of biomedical optics because it is highly potential for high-resolution cross-sectional imaging underneath the surface of biological tissue[1]. A lot of intensive works are now made for improvement of image quality and resolution, and combination OCT and endoscopes[2]. In particular, we demonstrated ultra high-resolution OCT and optical tomography of the geometrical dimensions suitable for clinical diagnoses[3-5].

OCT utilizes the low coherence interferometry to obtain a cross-sectional images in the way that is analogous to ultrasound echo imaging (B-mode operation). The spatial resolution of OCT images along the tissue depth, therefore, is determined by the coherence length of the light source itself. In the case where a super-luminescent diode (SLD) is used as the light source of the interferometer, the resolution of 10 to 20μm is obtained. Essentially, OCT is 2-D mapping of reflection of light with less influence of scattering in biological tissue; in other words, only the ballistic photons are selectively detected in OCT. Accordingly, the conventional Michelson interferometer can be used even for optical imaging of biological tissue which is strong scattering media. Thus, a miniaturized and vibration-free Michelson interferometer is easily realized by the use of a single-mode fiber directional coupler. The SM fiber is also very effective for selective detection of an extremely small amount of ballistic photons from scattered light. This means a practical OCT system is easily realized by the help of advanced optoelectronic technologies, and OCT is thus very promising for a variety of diagnoses in ophthalmology, dermatology, gastrointestinal surgery, etc.

At present, further technical development of OCT is now required to meet actual clinical diagnoses,

including improvement of the penetration depth of light and the spatial resolution. The penetration depth of light is mainly determined by scattering of biological tissue. Scattering, as a matter of course, is strongly dependent upon the light wavelength. In this paper, the imaging characteristics of OCT are presented for two specific wavelength of 0.8 and 1.3μm in the fiber-optic low coherence interferometers.

2. 0.8 and 1.3 μm fiber-optic OCT systems

Two different SLDs are commercially available for the wavelength of 0.8 and 1.3μm. In the past two decades, the 0.8-μm light has been used for measurement of concentration of oxidized and deoxidized Hb of blood. In contrast, the 1.3 or 1.5μm light is not popular in biomedical optics, but is the important wavelength range where optoelectronic components/devices are well developed. The fiber-optic OCT system is shown in Fig. 1, where the center wavelength of SLDs are 0.85 and 1.308μm with the spectrum bandwidth of 15 and 42nm, respectively. The output power of SLD is nearly 3mW. First of all, the coherence length were measured to be 15.2 and 18.0μm for 0.8 and 1.3μm SLDs, respectively. Both a reference mirror and a focusing lens are loaded on translation stages with an accuracy of 0.1 or 1μm. For optical imaging, the reference mirror is scanned at 10mm/s or more for each step of movement of the lens.

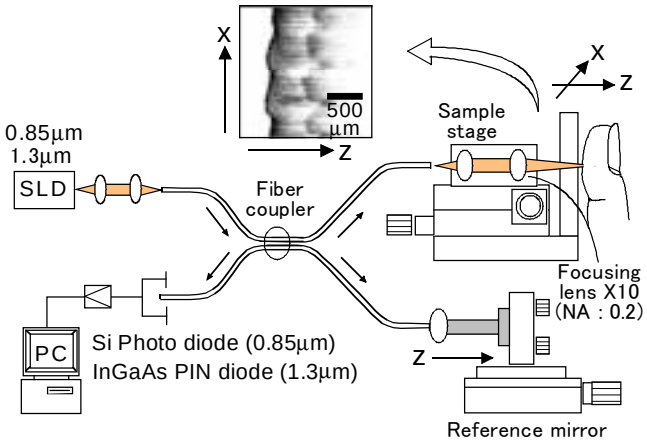
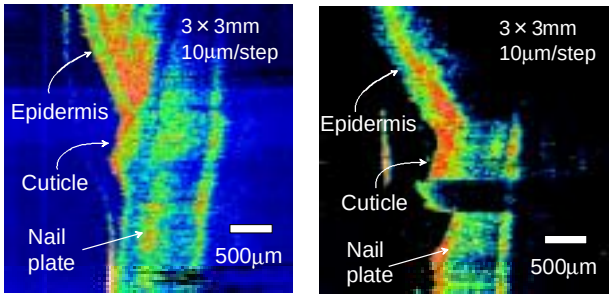


Fig. 1 System configuration of the fiber-optic OCT.

The use of fiber-optic interferometers is very effective for selective detection of ballistic photons, as shown in Fig 2, where 0.8μm bulk-optic and fiber-optic OCT images of human nail are compared. Obviously, the fiber optics can provide a higher contrast image with less speckle noise.



(a) (b)

Fig. 2 Comparison of in vivo OCT images of a human nail. (a) a bulk-optic OCT image. (b) a fiber-optic OCT image.

3. Evaluation of the maximum penetration depth

The maximum penetration depth of light, d_{max}, was evaluated using chicken tissue sandwiched by glass plates. In general, the detected light intensity I is expressed by $I = I_0 \exp(-2\mu_t d)$ and $\mu_t = \mu_a + \mu_s$,

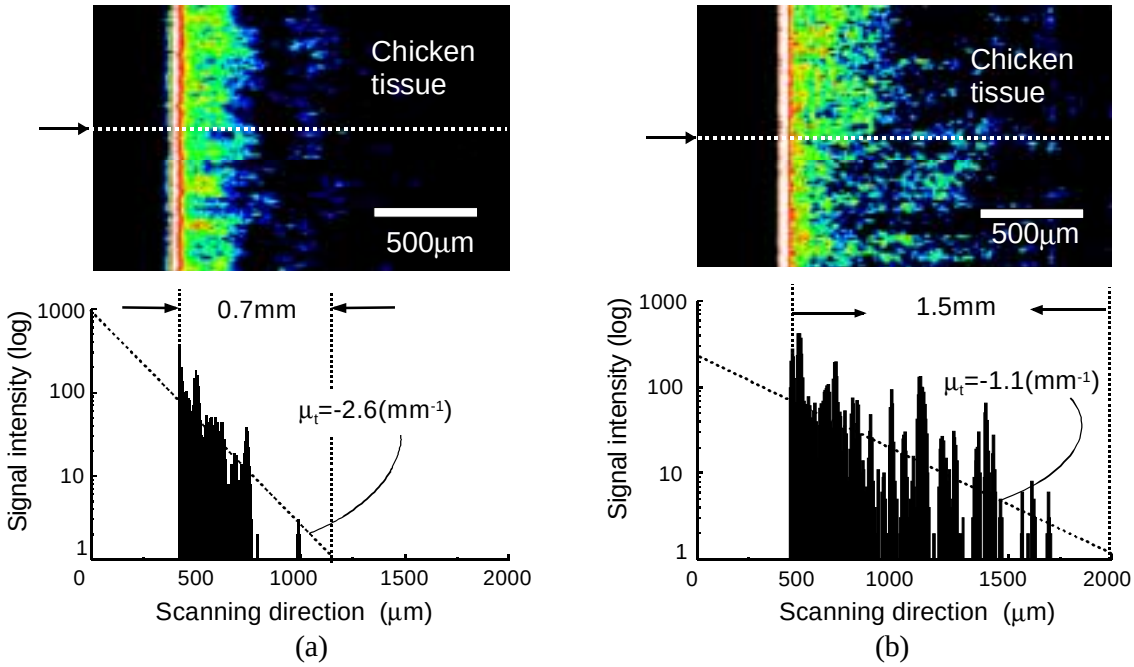


Fig. 3 In vitro OCT images of chiken tissue and the axial reflection profile along the raster. (a) 0.85μm OCT image, (b)1.3μm OCT image.

where I_0 is the incident light intensity, d is the penetration depth of light, μ_a and μ_s are absorption and scattering coefficients, respectively. In biological tissue, $\mu_s >> \mu_a$ is always satisfied. On the basis of the discussion described here, μ_t and d_{max} are evaluated from the 0.8μm and 1.3μm OCT images of Figs. 3 (a) and (b), which are false color images of the logarithmic light intensity level. μ_t is determined as the slope of a fitting line for the light intensity profile along a raster, as shown by the dotted line in Figs. 3 (a) and (b). d_{max} is also given by the distance from the tissue surface to the crossing point of the fitting line and the noise level. Consequently, we obtained μ_t = 2.6 and 1.1mm^{-1} and d_{max} = 0.7 and 1.5mm for 0.8 and 1.3μm OCT images, respectively. It is thus found out that d_{max} for 1.3μm OCT becomes almost twice as long as that for 0.8μm OCT. Such a drastic improvement of d_{max} is due to remarkable reduction of μ_t by the use of 1.3μm SLD light. In our experiment, the ratio of $\mu_{t\,1.3\mu m}$ and $\mu_{t\,0.8\mu m}$ is obtained to be 0.42 which is in good agreement with the estimated value from Ref. 6.

As an example of in vivo imaging, we present here 0.8 and 1.3μm OCT images of human nail, as shown in Fig. 4. The 0.8μm light gives us the cross sectional image of only a nail plate, while the 1.3μm light penetrates through to soft tissue behind a nail plate.

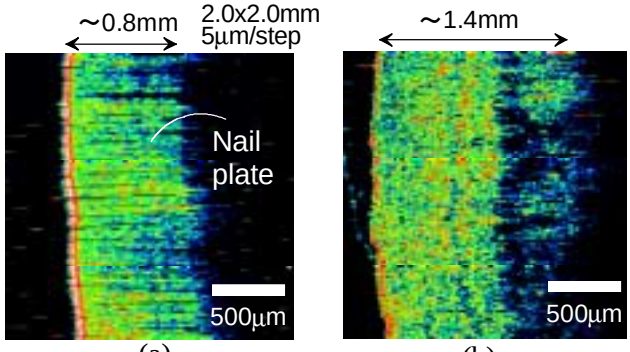
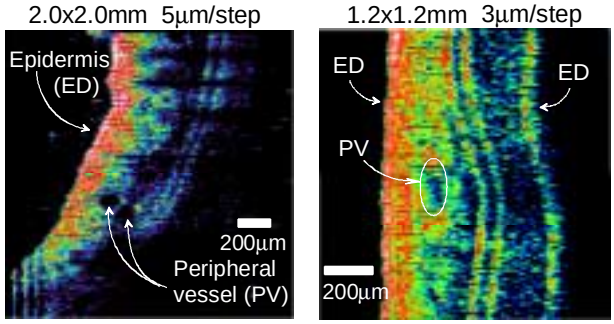


Fig. 4 Comparison of in vivo OCT images of a human nail. (a) 0.85μm OCT image, (b) 1.3μm OCT image.

4. Other imaging properties

The 1.3μm OCT has the advantage of deep penetration of light, but this type of OCT is not always suitable for cross-sectional imaging of biological tissue. In the near future, for example, the 2-D mapping of concentration of oxidized and deoxidized Hb should be required for function analyses of biological tissue. Such a function imaging is possible only by the 0.8μm OCT, in which at least two SLDs are necessary for different-wavelength imaging around 0.81μm.



(a) (b)

Fig. 5 In vivo OCT images of an earlobe of nude mouse. (a) 0.85μm OCT image, (b) 1.3μm OCT image.

In the 0.8μm OCT, the reflection light intensity reduces steeply along the depth due to a large scattering coefficient. This fact is not always shortcoming for the cross-sectional imaging. In the case of imaging of micro layer tissue near the surface, steeper reduction of reflection light leads to higher contrast of OCT images. We here present 0.8 and 1.3μm OCT images of an earlobe of nude mouse, as shown in Fig. 5. It can be found that the 0.8μm SLD can image more clearly tissue layers including peripheral blood vessels.

5. Conclusion

We compared two OCT images for 0.8μm and 1.3μm by use of the fiber-optic low coherence interferometers. The maximum penetration depth and the total attenuation coefficient of two light sources were evaluated using chicken tissue. Our experiment suggested that the maximum penetration depth for 1.3μm OCT is almost twice as long as that for 0.8μm OCT. On the contrary, it can be found that the 0.8μm OCT can image micro tissue structures near the surface with higher contrast. Although two properties, long penetration depth and high image contrast, are inconsistent with each other. Our effort is now directed at improvement of image contrast of 1.3μm OCT by the use of the light focusing effect of an objective.

References

1. D. Huang, E. A. Swanson, C. P. Lin, J. S. Schuman, W. G. Stinson, W. Chang, M. R. Hee, T. Flotte, K.Gregory, C. A. Puliafito and J. G. Fujimoto, Sciences **254**, pp. 1178-1181 (1991).
2. G. J. Tearney, S. A. Boppart, B. E. Bouma, M. E. Brezinski, N. J. Weissman, J. F. Southern and J. G. Fujimoto, Opt. Lett. **21**, pp. 543-545 (1996).
3. M. Haruna, K. Yoden, M. Ohmi, Y. Ohnishi and N. Kunizawa, OSA Biomed. Topical Meeting, **MB5**, Tech. Dig. pp.194-196, Maiami ,FL (2000).
4. K. Yoden, M. Ohmi, Y. Ohnishi, N. Kunizawa and M. Haruna, Opt. Rev. **7**, pp.402-405 (2000).
5. M. Haruna, R. Konoshita, M. Ohmi, N. Kunizawa and M. Miyachi, Proc. SPIE (Int'l Soc. for Opt. Eng.), Biomedical Optics, Vol. 4257 (2001), to be published.
6. P. Parsa, S. L. Jacques and N. S. Nishioka, Appl. Opt. **28**, pp. 2325-2330 (1989).